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Photonic crystal optical biosensor incorporating structured low-index porous dielectric

Ian D. Block [1], Leo L. Chan [1], Brian T. Cunningham *

Nano Sensors Group, University of Illinois at Urbana-Champaign, USA

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Abstract

The sensitivity of a photonic crystal optical biosensor is greatly enhanced through the incorporation of low refractive index porous dielectric material into the device structure. In this work, computer models are used to predict the reflectance spectra and sensitivity performance of a one-dimensional photonic crystal biosensor. A manufacturable replication method is demonstrated that can produce a low-index dielectric periodic surface structure with a 550 nm period over large surface areas. The sensitivity of porous glass biosensors is characterized and compared with sensors incorporating non-porous polymer material. Results for detection of proteins, polymer layers, and bulk liquids indicate up to a four-fold sensitivity increase.

Keywords: Optical biosensor; Low-k dielectric; Micro-replication; FDTD; Direct assay; Photonic crystal

1. Introduction

Label-free optical biosensors based upon surface structured photonic crystals have recently been demonstrated as a highly sensitive method for performing a wide variety of biochemical and cell-based assays [1]. The device structure is designed to reflect only a narrow band of wavelengths when illuminated with white light at normal incidence, where positive shifts of the reflected peak wavelength value (PWV) indicate the adsorption of detected material on the sensor surface [2]. By spatially confining incident photons at the resonant wavelength, a high optical field is generated at the sensor surface that extends a short distance into a test sample, much like an evanescent field. The high degree of spatial confinement of resonant photons within the device structure leads to a strong interaction between the structure and adsorbed biomaterial, and to the ability to perform high resolution imaging of protein and cell attachment [3].

Previously, photonic crystal optical biosensors have been fabricated from continuous sheets of plastic film using a process in which the periodic surface structure is replicated from a silicon master wafer using a UV-cured polymer material [4]. This patterned polymer is subsequently coated with a high refractive index TiO_2 layer that is generally thinner than the height of the surface structure. Such devices have been demonstrated for a wide variety of biochemical and cell-based assays, with a mass density sensitivity resolution less than 0.1 pg/mm^2 and a large dynamic range enabling single cell detection [5]. In general, optimization of the device sensitivity requires increasing the interaction of the electromagnetic field intensity distribution with the biological material deposited atop the photonic crystal surface. Therefore, selection of optical materials and design of the surface structure topology should be aimed at extending the electromagnetic field profile from the interior regions of the photonic crystal (where they cannot interact with adsorbed material) to the region adjacent to the photonic crystal that includes the liquid test sample. In this work, we demonstrate that the substitution of an extremely low refractive index material for the surface structure within the photonic crystal biosensor has the desired effect of substantially increasing detection sensitivity.

We used rigorous coupled wave analysis (RCWA) and finite difference time domain (FDTD) simulations to predict the resonant wavelength and bulk refractive index sensitivity of a one-dimensional surface photonic crystal biosensor. The device incorporates a low-index ($n = 1.17$) nanoporous dielectric surface structure in place of the polymer ($n = 1.39$) surface structure

* Corresponding author. Tel.: +1 217 265 6291.
E-mail address: bcunning@uiuc.edu (B.T. Cunningham).
[1] These authors contributed equally to this work.

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reported previously. We use a soft contact embossing method to create a surface-structured low-index porous film on glass substrates with a depth and period that are identical to the previous polymer structures to enable a side-by-side sensitivity comparison. The sensitivity of porous glass biosensors is compared to nonporous polymer biosensors through methods that characterize sensitivity to bulk refractive index and surface-adsorbed material. Finally, a protein binding assay comparison is performed to demonstrate sensor stability and the ability to functionalize the device for selective detection.

2. Materials and methods

2.1. Computer simulation

The polymer and porous glass sensors were modeled and simulated using two software packages. First, a 2-D diffraction grating analysis tool (GSOLVER) employing the RCWA algorithm provides a quick and simple method for initial sensor modeling. Second, FDTD (Lumerical) provides a much more versatile and powerful tool that can calculate any field component at any temporal or spectral location for an arbitrary optical device illuminated by an arbitrary source [6]. FDTD was used to verify RCWA results and to gain deeper insight into the effects of modifying the sensor structure.

2.2. Sensor fabrication

In the proposed device, we incorporate a sol–gel derived low-index nanoporous silica thin-film [7] in place of the UV-cured epoxy used in previous designs. Since the low-index material cures by heat rather than UV exposure, it was necessary to develop a new fabrication process. We desired to retain a low-cost imprinting method, though it was obvious that a plastic substrate could not sustain the requisite high temperatures for porous glass annealing. One possible approach to sol–gel glass imprinting was to use a polydimethylsiloxane (PDMS) mold and a glass substrate [8].

The sub-wavelength grating structure of the low-k biosensor was fabricated using a combination of lithography, molding, and imprinting processes. The process begins with production of a silicon "master" wafer, that is patterned by deep-UV photolithography with a linear grating photoresist pattern of parallel 275 nm width photoresist lines separated by 275 nm width spaces. The photoresist is used as a mask for reactive-ion etching of ~170 nm deep trenches into the silicon wafer. After etching, the photoresist is removed, and the silicon wafer contains a positive image of the surface structure desired in the finished sensor. Sylgard 184 PDMS (Dow) "daughter" molds are then replicated from the silicon master wafer. The liquid PDMS is poured into a rectangular metal frame placed on top of the silicon master wafer and then cured at 90 °C for 24 h. To facilitate release of the cured PDMS mold from the silicon wafer, the wafer was surface treated with a release layer of dimethyldichlorosilane (Repel Silane, Amersham Biosciences) [9]. The PDMS replicas are then used to imprint a thin-film of uncured Nanoglass® (Honeywell Electronic Materials), a low-index sol–gel glass, spun-on

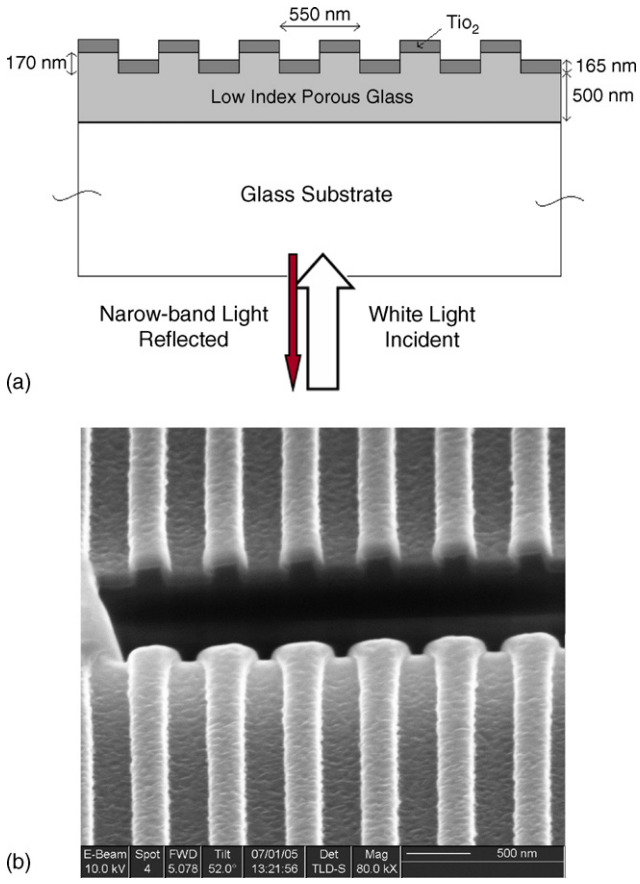

Fig. 1. Cross-section schematic (a) and scanning electron micrograph (SEM) (b) of porous glass photonic crystal sensor. Surface roughness visible in SEM is due to a carbon coating atop the TiO_2 film used to enhance image quality.

to a glass substrate. Once the low-index dielectric becomes rigid, the flexible PDMS mold is removed and the sol–gel glass is fully cured by further baking. The sensor structure is completed by evaporating 165 nm of TiO_2 onto the patterned surface. A subsequent surface treatment with dimethyldichlorosilane encourages bio-adsorption and promotes sensor stability. A schematic illustration and scanning electron micrograph of the device cross-section are shown in Fig. 1(a and b).

The polymer structure is similar to that described in a previous publication [4]. It contains a periodic surface structure of UV-cured polymer that is subsequently overcoated with a layer of TiO_2 deposited by a sputtering process. The polymer periodic surface structure is also formed by a replica molding process, in which an 8-in. diameter silicon "master" wafer is used as a mold. A thin layer of UV-curable polymer is squeezed between a flexible transparent plastic substrate film and the silicon wafer. The polymer is allowed to flow into the silicon mold before it is rapidly cured to a solid by exposure to UV light. After curing, the replica (attached to the plastic film) is peeled away from the silicon mold, so the silicon may be used for additional replicas. Because the same silicon "master" is used to produce the polymer and porous glass devices, both structures have a 550 nm period and 170 nm imprint depth. The grating period and grating depth were verified by scanning electron microscopy. The